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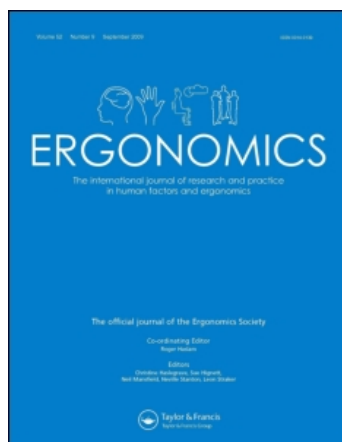
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When is a lifting movement too asymmetric to identify low-back loading by 2-D analysis?

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In ergonomics research, two-dimensional (2-D) biomechanical models are often used to study the mechanical loading of the low back in lifting movements. When lifting movements are asymmetric, errors of unknown size may be introduced in a 2-D analysis. In the current study, an estimation of these errors was made by comparing the outcome of a 2-D analysis to the results of a recently developed and validated 3-D model. Four subjects made two repetitions of five lifting movements, differing in the amount of asymmetry. The results showed a significant underestimation of the peak torque by 20, 36 and 61% when the initial position of a box was rotated 30, 60 and 90° with respect to the sagittal plane of the subject. The main cause of this underestimation was a pelvic twist, resulting in an erroneous projection of a pelvic marker on to the sagittal plane due to pelvic twist. It is suggested that from 30° box rotation a 2-D analysis may easily lead to wrong conclusions when it is used to study asymmetric lifting.

1. Introduction

The mechanical load on the low-back in lifting activities is frequently studied by use of linked-segment models. In such an approach, net reactive forces and torques at, for instance, the lumbosacral joint (L5-S1) are calculated by an inverse dynamic analysis, starting either at the hands or at the feet. Most of these models are two-dimensional (2-D) and hence they are restricted to the analysis of movements in the sagittal plane (Chaffin 1969, Freivalds *et al.* 1984, Bush-Joseph *et al.* 1988, de Looze *et al.* 1992). However, most lifting movements in daily occupational life are not symmetric.

It is obvious that errors are introduced when 2-D models are used to analyse asymmetrical lifting movements. When tasks are compared that differ in the amount of asymmetry, such errors may result in wrong decisions in ergonomics practice. Introduction of 3-D models into occupational research might solve this problem at the cost of more complex measurements and data analysis. In order to decide in what situations it is necessary to accept these costs, the question that must be answered is what is the amount of asymmetry that leads to unacceptable errors in 2-D models? The answer to this question can be found when the outcome of a 2-D model is compared to the outcome of a 3-D model.

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Although several 3-D linked-segment models have been developed previously (Kromodihardjo and Mital 1987, Frigo 1990, Gagnon *et al.* 1993), no attempts were made in these to compare the results to the outcome of 2-D models.

The current study was designed to make such a comparison. In various asymmetrical lifting conditions, torques at the L5-S1 joint, calculated with a 2-D model (de Looze *et al.* 1992), were compared to the torques calculated by a recently developed and validated 3-D model (Kingma *et al.* 1996a).

2. Methods

Four healthy young males (weight 70.7 ± 5.6 kg, height 182.7 ± 11.4 cm) participated in this experiment after signing an informed consent form. One symmetrical and four asymmetrical conditions were created by placing a 15.7 kg box in different positions. The first location was right in front of the feet at a distance of 5 cm, so that a symmetrical lifting movement could be performed (0° rotation condition). The second to fifth locations were created by rotating the box to the right side of the subject by respectively 10, 30, 60 and 90° around a vertical axis between both ankles. The position of the feet was kept constant. The shortest distance from the box to the right foot was kept constant at 5 cm. A switch on the box generated a synchronization pulse at lift-off. Prior to the experiment the subjects were free to choose a comfortable initial body posture. In this posture the height of the left hip was measured (97 ± 3.8 cm). This height was kept constant in all conditions. The subjects were asked to hold the handles of the box gently in the starting position, and after counting down, to lift the box to knuckle height in a symmetrical upright standing position. A metronome was used to control the speed of the lifting movements.

Each lifting movement was repeated once, so that a total of 10 lifting movements was performed by all subjects. During the lifting movements, the positions of reflective markers were recorded at 60 Hz using a 3-D automatic video-based motion recording system (VICON, Oxford Metrics, Oxford). Ground reaction forces were recorded simultaneously by two force-platforms (Kistler, 9218B Winterthur, Switzerland) and, after analogue low-pass filtering at a cut-off frequency of 30 Hz, digitized at 60 samples/s.

For both the 2-D and 3-D linked segment model, segment masses and moments of inertia were derived with the aid of anthropometric measurements and regression equations described by McConville *et al.* (1980).

A 2-D dynamic linked-segment model, using inverse dynamics (de Looze *et al.* 1992), was used to calculate sagittal plane torques at the L5-S1 joint. This model used the sagittal plane coordinates of reflective markers attached to landmarks at the fifth metatarsal joint, the lateral malleolus, the lateral femoral epicondyle, the greater trochanter and the L5-S1 joint on the left side of the body. Segment angles were calculated as the angle between the line connecting two successive markers and the forward directed horizontal. Joint positions were represented by the markers. Centres of mass were calculated as a ratio of the distance between two successive markers. Segment linear and angular accelerations were obtained from the time histories of, respectively, the segment centres of gravity and the segment angles, by double differentiation using a Lanczos five-point differentiator.

The 3-D linked-segment model, again using inverse dynamics, calculated the torques at the L5-S1 joint in all three planes of movement. This model has been described in Kingma *et al.* (1996a). In short: to both feet, lower legs, upper legs and to the pelvis a brace constructed of 5 mm thick thermoplastic material (Orfit) was

attached. The braces could be adapted to individual segment contours by briefly heating them. To each brace, five spherical markers (10 mm in diameter) were mounted using rigid thread (3 mm in diameter) of varying length. Prior to the experiment a 'calibration recording' was made for each body segment. For this purpose additional reflective markers were mounted to the segment at relevant anatomical landmarks in order to allow reconstruction of an anatomical axis system. During calibration, the position of these markers was recorded simultaneously with the markers on the braces. After this recording the markers on the anatomical landmarks were removed.

The markers on the anatomical landmarks were used to reconstruct an anatomical axis system and to calculate the inertia tensor, the centre of gravity position and joint centre position of the segments at the time of the calibration recording. Additionally, the positions of the five markers on each brace, during both the segment calibration recording and during time instant i of a lifting movement, were used to calculate the transformation of each segment from calibration position to the position at instant of time i of the lifting movement. The transformation (a rotation matrix R_j , translation vector \mathbf{r}_j and mean brace marker position \mathbf{r}_j) was calculated with a least squares algorithm developed by Veldpaus *et al.* (1988). Subsequently, the transformation was applied to parameters calculated for the segment calibration position (i.e. to the anatomical axis system, the inertia tensor, the centre of gravity position and joint centre position). In this way, the kinematic input for the 3-D model was generated.

Marker positions of the 2-D model and segment centre of gravity positions of the 3-D model were digitally filtered using a fourth order Butterworth filter with zero phase lag at an effective cut-off frequency of 5 Hz. Segment linear accelerations were obtained from the time histories of the segment centres of gravity by double differentiation using a Lanczos five-point differentiator. The same differentiator was applied once to the time histories of the nine elements of the inertia tensor in order to obtain the first derivative of the inertia tensor. Angular speeds and angular accelerations of the segments were calculated from the rotation matrices \mathbf{R}_j according to Berme *et al.* (1990).

The inverse dynamic process started at both feet, using the data described above and the data from both forceplates. A formulation of the equations of motion in the global axis system was used. At each instant of time the body segment is subject to the following two equations of motion:

$$\sum_{k=1}^P \mathbf{F}_k + m\mathbf{g} = m\mathbf{a}, \quad (1)$$

$$\sum_{k=1}^P ((\mathbf{r}_{r,k} - \mathbf{r}_{com}) \times \mathbf{F}_k) + \sum_{l=1}^q \mathbf{M}_l = d(\mathbf{I}\omega)/dt = d(\mathbf{I})/dt \omega + \mathbf{I}\alpha, \quad (2)$$

where \mathbf{F}_k are all p external and intersegmental forces k , applied at the body segment; m is the segment mass and \mathbf{I} is the inertia tensor; \mathbf{g} is the gravity vector; \mathbf{a} is the segment linear acceleration; ω and α are, respectively, the angular speed and angular acceleration of the segment; $\mathbf{r}_{r,k}$ is the point of application of force k , i.e. a joint centre or the point of application of an external force; \times is a vector product; \mathbf{M}_l are all q torques l , applied at the body segment.

The global axis 3-D torque at the L5-S1 joint that was calculated in this way, was projected on the pelvic anatomical axis system in order to improve anatomical interpretation of the torques.

A time period (from 0.583 s before the synchronization pulse until 1.083 s after the synchronization pulse) was selected for further analysis. This period was selected in order to (1) include all lifting movements completely, and (2) avoid influence of variations in sampling time during upright standing between trials on mean torques. Mean and peak values of the resulting torques were calculated for the sagittal plane torque (2-D model), flexion-extension torque (3-D model) and the total torque, i.e. the

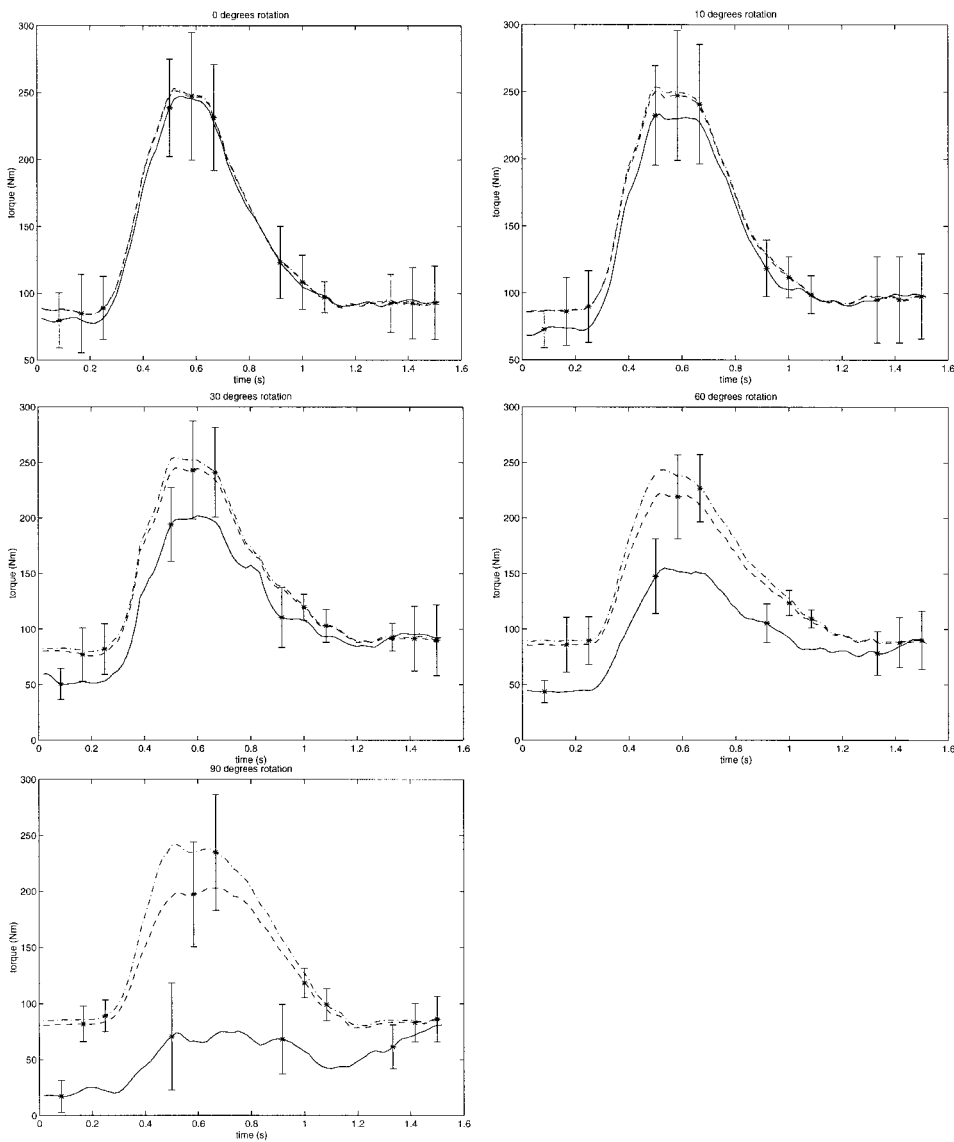


Figure 1. Averaged curves over subjects for the 2-D model sagittal plane torque (solid line), the 3-D model flexion-extension torque (dashed line) and the 3-D model total torque (dash dotted line), in all five rotation conditions. Error bars indicate 1 SD. *Indicates intersections between the error bars and the curves to which they belong.

square root of the sum of the three squared torque components (3-D model). After averaging between the two trials within each condition and subject, the difference between the 3-D model flexion-extension peak torque and the 2-D model sagittal plane peak torque was tested using an ANOVA with the rotation condition and subject as factors. This was repeated for the mean torques. The same procedure was applied to the difference between the 3-D model total torque and the 2-D model sagittal plane torque. In case of a significant main effect of rotation condition, one-sided Dunnett *post-hoc* tests were used to test if the difference between 3-D model and 2-D model torques in each rotation condition was higher as compared to the sagittally symmetrical lifting movements. A *p*-value < 0.05 was considered to be significant.

3. Results

Figure 1 shows the averaged curves for the 2-D model sagittal plane torque, the 3-D model flexion-extension torque and the 3-D model total torque, in all five box rotation conditions. The figure shows that in the symmetrical lifting condition there are no substantial differences between 2-D and 3-D torque estimates. In contrast, an increasing underestimation of the torque calculated by the 2-D model was found with increasing box rotation. The ANOVA indicated main effects of the rotation condition on the difference between 3-D model and 2-D model torques. This was the case for mean as well as peak torques and for the 3-D model total torque as well as the 3-D model flexion-extension torque difference with the 2-D model sagittal plane torque (table 1).

For the sagittally symmetrical lifting movements the difference between the 3-D model and 2-D model torque estimates amounted to about 1 to 3%. For mean as well as peak torques the increase of the difference between the 3-D model and 2-D model torque estimates in the 10° rotation condition as compared to the sagittally symmetrical lifting movements was not significant (tables 2 and 3). For 30° or more box rotation all differences between the 3-D model and 2-D model torque estimates were significantly higher as compared to the sagittally symmetrical lifting movements. For the 3-D model flexion-extension minus 2-D model sagittal plane mean torque this difference was 18, 33 and 62% (table 2) of the 3-D model flexion extension torque. For the peak torques the differences were 20, 36 and 61% (table 3) in the 30, 60 and 90° rotation condition respectively. The difference between the 3-D model flexion-extension and the 2-D model sagittal plane torque was similar at 16, 29 and 58% for mean torques (table 2) and at 18, 30 and 53% for peak torques (table 3) for the 30, 60 and 90° rotation condition respectively.

Table 1. ANOVA results for the effects of rotation of the box with respect to the sagittal plane and of subject effects on the difference between 3-D model and 2-D model estimates of the (mean and peak) torque estimates at the lumbosacral joint.

	Rotation effect		Subject effect	
	<i>F</i> (4,12)	<i>p</i>	<i>F</i> (3,12)	<i>p</i>
Mean 3-D flexion extension– 2-D	33.7	< 0.001	12.6	0.001
Mean 3-D total– 2-D	47.0	< 0.001	13.0	< 0.001
Peak 3-D flexion extension– 2-D	28.2	< 0.001	9.2	0.002
Peak 3-D total– 3-D	25.4	< 0.001	4.7	0.021

Table 2. Mean torques (averaged over subjects), estimated by the 2-D model (column 2), the 3-D model flexion-extension component (column 4) and the 3-D model total torque (column 7). *Post-hoc* tests were applied to find out for each rotation condition if the difference between the 3-D model and the 2-D model torques was significantly higher as compared to the sagittally symmetrical lifting movement. Test results (*p*-values and the standard error) are given in column 6 for the 3-D model flexion-extension torque and in the last column for the 3-D model total torque.

Rotation	2-D sagittal plane torque		3-D flexion extension torque		Rotation effect on difference with 2-D (<i>p</i> -value)	3-D total torque		Rotation effect on difference with 2-D (<i>p</i> -value)
	(Nm)	(SD)	(Nm)	(SD)		(Nm)	(SD)	
0	132	(17)	135	(20)		136	(20)	
10	129	(20)	138	(20)	0.447	139	(21)	0.421
30	112	(12)	133	(16)	0.033	137	(15)	< 0.016
60	93	(20)	131	(17)	< 0.001	138	(16)	< 0.001
90	53	(26)	125	(18)	< 0.001	138	(19)	< 0.001
SE					6.7			6.8

Table 3. Peak torques (averaged over subjects), estimated by the 2-D model (column 2), the 3-D model flexion-extension component (column 4) and the 3-D model total torque (column 7), *Post-hoc* tests were applied to find out for each rotation condition if the difference between the 3-D model and the 2-D model torques was significantly higher as compared to the sagittally symmetrical lifting movement. Test results (*p*-values and the standard error) are given in column 6 for the 3-D model flexion-extension torque and in the last column for the 3-D model total torque.

Rotation	2-D sagittal plane torque		3-D flexion extension torque		Rotation effect on difference with 2-D (<i>p</i> -value)	3-D total torque		Rotation effect on difference with 2-D (<i>p</i> -value)
	(Nm)	(SD)	(Nm)	(SD)		(Nm)	(SD)	
0	252	(52)	255	(50)		256	(50)	
10	240	(43)	256	(45)	0.340	259	(46)	0.448
30	206	(30)	251	(44)	0.006	259	(42)	0.025
60	161	(35)	229	(40)	< 0.001	250	(38)	0.001
90	98	(37)	209	(46)	< 0.001	254	(69)	< 0.001
SE					11.5			17.2

4. Discussion

Since human movements are never completely 2-D, it is quite obvious that the estimation of joint loads by a 2-D model will result in some errors. Besides skin movement artefacts and errors in the estimation of segment inertial parameters (Kingma *et al.* 1996b), these errors are caused by the neglect of other than sagittal plane torque components, leading to an underestimation of the total torque. However, the flexion-extension torque can be a good estimator of the total torque, as long as the flexion-extension torque exceeds the other components by far, since the total torque is calculated by taking the square root of the summed squares of the three components.

Figure 1 shows that this is the case, at least for the 0, 10 and 30° rotation condition, since the 3-D model flexion-extension torque is close to the 3-D model total torque.

The current study shows that the 2-D model already tended to be in error when the box to be lifted is rotated 10° out of the sagittal plane (figure 1), although the *post-hoc* tests showed no significant effect on mean and peak torques. While the size of the errors (7% for mean as well as peak torques) may be judged as acceptable in 10° box rotation, the errors become quite large (20% for peak torques) when the box rotation is 30° or more. If measurements during a lifting task in occupational practice are used to compare torques to some standard, or if a population at risk is calculated, errors of the order of 20% may often be judged to be unacceptable. If torques are used to compare tasks, a 2-D analysis may result in erroneous conclusions if the tasks differ in the amount of asymmetry.

The most important problem in the application of 2-D models is the projection of markers on the sagittal plane. The placement of markers on one side of the body, which is the normal procedure in a 2-D analysis of human movement, can enhance the errors. In the current study, the 2-D model markers were placed on the left side

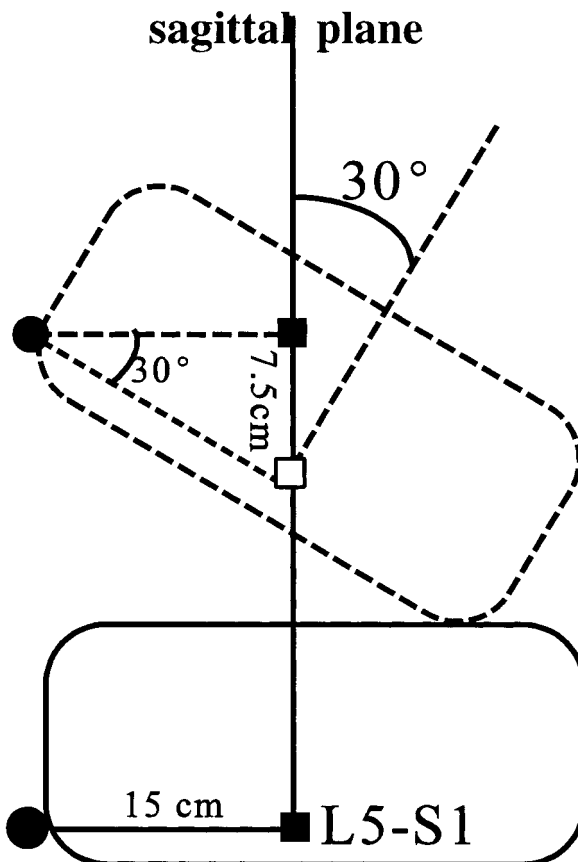


Figure 2. Top view of a pelvis showing the projection error that a 2-D model makes when the pelvis is rotated.

of the body whereas the subjects rotated to the right. This resulted in an estimation of the L5-S1 joint centre too far forward, causing an underestimation of the flexion-extension torque. Figure 2 shows an example of such a situation: the pelvis is rotated 30° to the right, leading to an error in the estimation of the L5-S1 joint of 7.5 cm. This results in large errors in torque estimates (de Looze *et al.* 1992). It should be realized that a rotation of the subject to the left would have caused a rotation backward of the 2-D L5-S1 marker with respect to the real L5-S1 joint centre. This would have resulted in an *overestimation* of the flexion-extension torque by the 2-D model of the same order of magnitude.

The projection error mentioned here may be solved in a relatively simple way by attaching a marker at the same location on the right side of the body. Subsequently, the L5-S1 joint centre position is calculated by averaging the left and right side marker. In this way, a 2-D model may still give a reasonable estimate of the flexion-extension torque when a box is rotated through 30° . However, estimating the location of markers on both sides of the body is only possible if more than one camera is used.

Furthermore, recent epidemiologic research shows that the extent of asymmetry in occupational lifting movements is associated with an increased risk of acute herniation of the intervertebral disc (Kelsey *et al.* 1984, Marras *et al.* 1995). This suggests that at least in part the aetiology is associated with asymmetry factors. Therefore, even if 2-D model errors in the estimation of flexion-extension torques are reduced, it may be important to use a 3-D model in order to quantify lateral flexing and twisting torques.

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References

- BERME, N., CAPOZZO, A. and MEGLAN, J. 1990, Rigid body mechanics as applied to human movement studies, in N. Berme and A. Capozzo (eds), *Biomechanics of Human Movement: Applications in Rehabilitation, Sports and Ergonomics* (Worthington, OH: Bertec), 89–107.
- BUSH-JOSEPH, C., SCHIPPLEIN, O., ANDERSSON, G. B. J. and ANDRIACCHI, T. P. 1988, Influence of dynamic factors on the lumbar spine movement in lifting, *Ergonomics*, **31**, 211–216.
- CHAFFIN, D. B. 1969, A computerized biomechanical model-development of and use in studying gross body actions, *Journal of Biomechanics*, **2**, 429–441.
- DE LOOZE, M. P., BUSSMANN, J. B. J., KINGMA, I. and TOUSSAINT, H. M. 1992, Validation of a dynamic linked segment model to calculate joint moments in lifting, *Clinical Biomechanics*, **7**, 161–169.
- FREIVALDS, A., CHAFFIN, D. B., GARG, A. and LEE, K. S. 1984, A dynamic biomechanical evaluation of lifting maximum acceptable loads, *Journal of Biomechanics*, **17**, 251–262.
- FRIGO, C. 1990, Three-dimensional model for studying the dynamic loads on the spine during lifting, *Clinical Biomechanics*, **5**, 143–152.
- GAGNON, M., PLAMONDON, A. and GRAVEL, D. 1993, Pivoting with the load. An alternative for protecting the back in asymmetric lifting, *Spine*, **18**, 1515–1524.
- KELSEY, J. L., GITHENS, P. B., WHITE, A. A., HOLFORD, T. R., WALTER, S. D., O'CONNOR, T., OSTFELD, A. M., WEIL, U., SOUTHWICK, W. O. and CALOGERO, J. A. 1984, An epidemiological study of listing and twisting on the job and risk for acute prolapsed lumbar intervertebral disc, *Journal of Orthopaedic Research*, **2**, 61–66.
- KINGMA, I., DE LOOZE, M. P., TOUSSAINT, H. M., KLINSMA, J. G. and BRUINEN, T. B. M. 1996a, Validation of a full body 3-D dynamic linked segment model, *Human Movement Science*, **15**, 833–860.

- KINGMA, I., TOUSSAINT, H. M., DE LOOZE, M. P., and VAN DIEËN, J. H. 1996b, Segment inertial parameter evaluation in two anthropometric models by application of a dynamic linked segment model, *Journal of Biomechanics*, **29**, 693–704.
- KROMODIHARDJO, S. and MITAL, A. 1987, Biomechanical analysis of manual lifting tasks, *Journal of Biomedical Engineering*, **109**, 132–138.
- MARRAS, W. S. M., LAVENDER, S. A., LEURGANS, S. E., FATHALLAH, F. A., FERGUSON, S. A., ALLREAD, W. G. and RAJULU, S. L. 1995, Biomechanical risk factors for occupational-related low-back disorders, *Ergonomics*, **38**, 377–410.
- McCONVILLE, J. T., CHURCHILL, T. D., KALEPS, I., CLAUSER, C. E. and CUZZI, J. 1980, Anthropometric relationships of body and body segment moments of inertia, Report no. AFAMRL-TR-80-119, Air Force Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Cleveland, OH.
- VELDPAUS, F. E., WOLTRING, H. J. and DORTMANS, L. J. M. G. 1988, At least squares algorithm for the equiform transformation from spatial marker coordinates, *Journal of Biomechanics*, **21**, 45–54.